PATENT APPLICATION

BY

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FOR

APPARATUS AND METHOD FOR PHASED SUBARRAY IMAGING

FIELD OF THE INVENTION

The present invention relates generally to an apparatus and method for phased subarray imaging, including image reconstruction, in an ultrasound imaging system. More particularly, by acquiring low-beam-rate images with a series of subarrays, interpolation and spectral modification, the present invention allows high-beam-rate images to be obtained while reducing the complexity of the front-end electronics in the ultrasound imaging system.

BACKGROUND OF THE INVENTION

Real-time medical ultrasound imaging has played an increasingly important role in the diagnosis and treatment of disease. Ultrasound imaging is used for routine diagnostic procedures in obstetrics, gynecology, cardiology, and gastroenterology. The vast majority of ultrasound systems in use today provide two-dimensional (2D) cross-sections of the anatomy. While other imaging modalities such as magnetic resonance imaging and x-ray computed tomography have provided three-dimensional (3D) images since their inception, only recently have 3D ultrasound imaging systems become commercially available. These systems have the potential to revolutionize medical imaging by providing 3D visualization of the anatomy and blood flow in real-time.

Conventional hardware and methods used for 2D ultrasound systems do not scale well to achieve similar 3D imaging systems. Modern 2D ultrasound scanners use a long 1D-transducer array having roughly 192 elements. Transducer array length and number of elements used is chosen based on several design parameters, including operating frequency and desired lateral resolution. An equivalent 3D imaging system capable of achieving similar resolutions in both azimuth and elevation would require a square 2D transducer array with 192 elements per side, or a total of 36,864 elements. A first challenge one faces when implementing such a system is fabricating the transducer array with reasonable yields.

A second challenge caused by a large channel count for a 3D ultrasound system is implementing the highly parallel front-end electronics required. Front-end hardware has become one of the most space- and power-consuming parts of a typical ultrasound imaging system. This is especially true since the advent of digital beamforming to vary transmit and receive directions and focal lengths, which has greatly reduced back-end hardware requirements. Unfortunately, the analog nature of the front-end hardware has not experienced an equal reduction in cost and size. High-end commercial ultrasound machines still house the analog and mixed-signal, front-end electronics within a base unit, requiring costly and bulky probe cables that contain dedicated coaxial transmission lines for each transducer element.

Modern 2D imaging systems require this complex set of front-end electronics because they typically use conventional full phased array (FPA) imaging, which requires that all array elements be simultaneously active during transmit and receive. See, for example, A. Macovski, "Medical Imaging Systems" (Prentice Hall, Englewood Cliffs, NJ, 1983). As shown in **Fig. 1**, in an FPA imaging system **100**, for every transducer element that is active for a given firing event (**110**, **112**, **114**, **116**, **118**, **120**, **122**, **124**), an independent front-end transmit (**126**, **128**, **130**, **132**, **134**, **136**, **138**, **140**) and receive (**142**, **144**, **146**, **148**, **150**, **152**, **154**, **156**) electronics channel must perform pulse generation, transmit/receive switching, amplification, filtering, time-gain compensation and digital-to-analog conversion in parallel. These electronics are the primary contributor to the bulk, cost, and power consumption of a typical ultrasound imaging system. In addition to high front-end hardware complexity, the large number of received signals required to form each beam causes a significant increase in transmit beamformer **158** and receive beamformer **160** complexity. The implementation of precision delay lines for beam steering also places a large burden on the beamforming hardware. Using all elements for transmit and receive results in the best image quality, improves signal-to-noise ratio (SNR) by maximizing total transmitted signal power, improves overall sensitivity for receiving echo signals, and has a very high frame rate since only one transmission or firing is required for each transmit direction. While electronic components continue to become smaller, faster, and cheaper, it is still not feasible to implement a full set of channels required for a 2D transducer array for 3D ultrasonic imaging.

The need to reduce the number of channels in a 3D imaging system has been recognized for some time, and several approaches have been presented in the art. One approach is the use of sparse arrays, which define a fixed subset of active elements that span a full aperture of the array. Different methods for choosing active elements include random and periodic distributions. Other array geometries intended to reduce the channel count include boundary arrays and a Mill's cross array. While these methods successfully reduce the

channel count of the system, they suffer from high side lobes (and thus poor contrast resolution) and low signal-to-noise ratio (SNR).

Alternative beamforming methods have also been suggested. As shown in **Fig. 2**, classical synthetic aperture (CSA) imaging techniques employing a single channel (or a few neighboring channels) for transmit and receive minimize the hardware complexity. In a CSA imaging system **200**, a transmit/receive controller **210** provides drive signals to an active element **216** via front-end transmit electronics **212** and a multiplexer **214** and receives received signals via the multiplexer **214** and front-end receive electronics **218**. See, for example, US Pat. No. 4,839,652. CSA was first used with linear arrays with reconstruction in the spatial domain, but has since been modified for use with circular arrays and frequency-domain reconstruction methods have also been developed. For a standard linear array method, a single processing channel is time-multiplexed across all transducer elements. Since only a single element is used for both transmit and receive, the complexity of the front-end electronics is kept to an absolute minimum; however, transmitted power and receive sensitivity are minimal and lead to low SNR. Each image pixel is reconstructed using all echo scans; time separation between scans leads to tissue motion artifacts. When used to construct images from an array with an element pitch equal to half of a minimum wavelength, CSA also suffers from high grating lobes. To avoid the grating lobes, element pitch is typically chosen to be a quarter of the minimum wavelength, but at the expense of reducing the physical aperture (and the related lateral resolution) by a factor of two for the same element count. CSA also requires multiple transmissions for each transmit direction and adversely impacts the frame rate.

In synthetic phased array (SPA) imaging with a single active element per data acquisition step, each image pixel is formed by coherent summation of signal contributions from every transmit/receive element combination. (SPA imaging is also shown in **Fig. 2**.) See, for example, US Pat. Nos. 4,586,135 and 5,465,722. SPA processing produces images with comparable resolution and SNR to the FPA images with lower front-end complexity. However, there is a significant increase in the number of transmissions for each image frame

with the usual adverse impact on the frame rate. In addition, the technique is limited by a limited transmit/receive power from a single active channel, which necessitates especially low electronic noise front-end electronics.

Array imaging techniques have continued to strike compromises between CSA and FPA, aiming to improve the SNR of CSA methods and reducing the number of channels required for FPA imaging. An early proposal for reducing the number of active channels in phased array imaging systems did so by transmitting on a single central portion of the array and receiving on a number of overlapping or adjacent subarrays. See, for example, US Pat. No. 4,553,437 and L.F. Nock et al., "Synthetic Receive Aperture Imaging with Phase Correction for Motion and for Tissue Inhomogeneities. I: Basic Principle," IEEE Trans. Ultrason., Ferroelect., Freq. Contr., vol. 39, pp. 489-95 (1992). Later developments improved the frame rate of subarray imaging by acquiring a subset of the beam lines and interpolating the others. See, for example, M. Karaman, "Ultrasonic Array Imaging Based on Spatial Interpolation," 3rd IEEE International Conference on Image Processing, pp. 745-748 (1996) and US Pat. No. 5,940,123. These methods, however, use 1D lateral interpolation filters and thus only produce successful results for relatively narrowband imaging. Recent proposals include transmitting from multiple elements to emulate a more powerful transmit element in SPA imaging, although a correction for motion and phase aberration would be required. A similar method proposes transmitting from five virtual elements and using the full aperture in receive in order to achieve the higher frame rates needed for 3D imaging with a 2D transducer array.

Real-time ultrasound imaging systems represent a tradeoff between front-end electronic complexity, image quality, SNR and frame rate. The proposals in the prior art do not successfully combine the advantages of CSA imaging in terms of reduced front-end complexity with the high quality image, high SNR and high frame rate associated with FPA imaging. Accordingly, there remains a need for a novel imaging system that combines the advantages of FPA and CSA imaging systems.

OBJECTS AND ADVANTAGES

In view of the above, it is a primary object of the present invention to provide an apparatus and method for phased subarray imaging, including image reconstruction, in an ultrasound imaging system. The phased subarray imaging of this invention provides a high-beam-rate image and allows a reduction in the front-end electronic complexity of the ultrasound imaging system.

These and numerous other objects and advantages of the present invention will become apparent upon reading the following description.

SUMMARY

The objects and advantages of the present invention are secured by an apparatus and method for phased subarray (PSA) imaging. An array of transducers is divided into a set of subarrays each having of multiple adjacent elements. Energy is transmitted with a transmit focal length from a subarray and complex responses to this energy are received by the subarray. The active subarray is multiplexed across the full array of transducers. Each subarray is fired multiple times to acquire Q_S beams, each defined by a direction in beam space and a plurality of receive focal lengths, that constitute a low-resolution subarray image with a low beam rate. The low-beam-rate subarray images are interpolated and spectrally modified to reconstruct high-beam-rate subarray images each having Q beams using at least one filter. The filter is varied for the subarrays. Appropriate weighting is applied to the high-beam-rate subarray images that are then combined to produce a high-beam-rate PSA image.

In an alternate embodiment, one subarray is used to transmit energy and another subarray is used to receive responses to this energy. The pair of transmit and receive subarrays are multiplexed across the full array of transducers.

In another embodiment, PSA imaging is performed for energy transmitted to a plurality of transmit focal lengths.

In another embodiment, the subarrays have a fixed number of adjacent elements. In another embodiment, the subarrays have a variable number of adjacent elements. In another embodiment, adjacent subarrays overlap one another and, therefore, have a number of common adjacent elements. In another embodiment, the overlap is fixed for all the subarrays. In another embodiment, the overlap between the subarrays is variable across the array. In yet another embodiment, there is no overlap between the subarrays, and different subarrays are used to transmit and receive energy.

The amount of overlap is chosen to ensure that an entire coarray (a measure of the spatial frequency content in the ultrasound imaging system) is covered by the subarrays and thus no image information is lost. There is a tradeoff between the number of subarrays, the amount of overlap and the frame rate. Reducing the number of subarrays and the amount of overlap results in a nonuniform coarray, which is not desireable for imaging. Therefore, in yet another embodiment, additional restoration filtering is applied to the high-beam-rate subarray images to restore the coarray for the PSA image to that of an FPA image.

In another embodiment, for sufficiently narrowband signals a filter for interpolation and spectral modification is a 1D or 2D filter. In another embodiment, for wideband signals the filter for interpolation and spectral modification is a 2D filter or a 3D filter.

In another embodiment, the subarrays have same number of said adjacent elements and the overlap of the subarrays is equal to half of the number of the adjacent elements in each of the subarrays and the filter for interpolation and spectral modification is a subarray-dependent bandpass filter with subarray-dependent gain or weighting.

In another embodiment, the filter for interpolation and spectral modification is varied for at least some of the subarrays as a function of the receive focal length.

In another embodiment, the upsampling and interpolation is varied for at least some of the subarrays.

In yet another embodiment, PSA imaging for at least some of the subarrays is repeated a plurality of times and the resulting high-beam-rate subarray images are averaged to improve the signal-to-noise ratio.

PSA imaging allows the number of front-end electronic channels to be reduced while maintaining high image quality as determined by a high-beam-rate and the signal-to-noise ratio. The quality of the final image is comparable to that achieved using FPA imaging for regions near the transmit focal length. The cost of PSA imaging is a reduction in the frame rate and SNR relative to FPA imaging. When the subarrays contain a fixed number of adjacent elements and neighboring subarrays overlap by less than half the number of adjacent elements in each subarray, the frame rate reduction is less than a factor of 2 for 2D imaging and less than a factor of 4 for 3D imaging.

BRIEF DESCRIPTION OF THE FIGURES

The objectives and advantages of the present invention will be understood by reading the following detailed description in conjunction with the drawings, in which:

Fig. 1 is a diagram illustrating a full phased array (FPA) imaging system as described in the prior art;

Fig. 2 is a diagram illustrating a classical synthetic aperture (CSA) imaging system and synthetic phased array (SPA) imaging as described in the prior art;

Fig. 3 is a diagram illustrating FPA imaging as described in the prior art;

Fig. 4 is a diagram illustrating phased subarray imaging (PSA) according to the present invention;

Fig. 5 is a diagram illustrating a 3D ultrasound imaging geometry including a 2D transducer array according to the present invention;

Fig. 6 is a block diagram illustrating the elements in an ultrasound imaging system according to the present invention;

Fig. 7 is a diagram illustrating a PSA imaging system according to the present invention;

Fig. 8 is a diagram illustrating restoration filtering in PSA imaging according to the present invention;

Fig. 9a is a diagram illustrating the comatrix for PSA imaging according to the present invention;

Fig. 9b is a diagram illustrating the comatrix for PSA imaging according to the present invention;

Fig. 10a is a diagram illustrating the spatial frequency response in PSA imaging according to the present invention;

Fig. 10b is a diagram illustrating the spatial frequency response in PSA imaging according to the present invention; and

Fig. 10c is a diagram illustrating the spatial frequency response in PSA imaging according to the present invention.

DETAILED DESCRIPTION OF THE EMBODIMENTS

Although the following detailed description contains many specifics for the purposes of illustration, anyone of ordinary skill in the art will readily appreciate that many variations and alterations to the following exemplary details are within the scope of the invention. Accordingly, the following preferred embodiment of the invention is set forth without any loss of generality to, and without imposing limitations upon, the claimed invention.

The basic geometry of a wideband pulse-echo sector-scanned 3D ultrasound imaging system is shown in **Fig. 5**. There is a 2D transducer array **500** having of N_1 elements **510** along a first lateral direction **512** and N_2 elements **514** along a second lateral direction **516**. Azimuth angle θ_1 **518** and elevation angle θ_2 **520** for radial direction **519** correspond to first lateral direction **512** and second lateral direction **516**. Axial direction **522** is normal to the plane defined by the first lateral direction **512** and the second lateral direction **516**.

The best resolution in the 3D ultrasound imaging system is obtained by minimizing the width of point spread function (PSF) in all three directions **512**, **516** and **522**. The width of the PSF in the first lateral direction **512** and the second lateral direction **516** is determined by the Fourier transform of an effective aperture function in these directions. Therefore, a wider effective aperture will lead to improved resolution along the corresponding lateral direction **512** and **516**. The lateral PSF and thus the lateral resolution are also improved as the temporal frequency is increased or the spatial wavelength is decreased (the temporal frequency and the spatial wavelength being related by the speed of sound). Because the lateral PSF is a function of angle, the lateral resolution in Cartesian coordinates will decrease as a function of radius. Axial width of the PSF, which determines axial resolution, is solely determined by a pulse function in the wideband pulse-echo ultrasound imaging system. The axial resolution therefore improves with shorter pulse lengths. For a fixed modulation frequency, a shorter carrier signal results in an increase of the bandwidth.

For a rectangular transmit aperture and receive aperture, the effective aperture, given by the convolution of the transmit aperture and the receive aperture, is triangular. The corresponding lateral PSF has a sinc-squared response. The central lobe of the PSF is proportional to wavelength λ divided by the effective aperture width D. Spatial frequency response or coarray is determined by the inverse Fourier transform of the lateral PSF. The coarray is simply a scaled version of the effective aperture. Like the PSF, the coarray characterizes the resolution of the ultrasound imaging system, i.e., how sensitive the system is to image features of different spatial frequencies.

Ultrasound array imaging systems involve several sampling schemes. The effective aperture is a sampled version of a continuous effective aperture reflecting the discrete spacing d_1 **524** and spacing d_2 **526** between elements in the array **500**. In the frequency domain, sampling results in a periodic repetition of the lateral PSF. In array imaging, the aliases are referred to as grating lobes and result in distortions of the image if they lie in the visible region. The visible region is defined as the range for azimuth angle θ_1 **518** and elevation angle θ_2 **520**

during transmission and reception between $-\pi/2$ and $\pi/2$ or, for $s_1=\sin\theta_1$ and $s_2=\sin\theta_2$ between -1 and 1. To avoid grating lobes in the visible region, the array **500** must be sampled with a period less than half of minimum wavelength (λ_{min}) in the pulse function. In addition, a finite number of beams or scan lines are determined over limited azimuth **518** and elevation **520** sector angles Θ_1 and Θ_2, with transmit and receive directions equally spaced is s_1 and s_2. For FPA, the minimum number of samples (beams) to avoid aliasing are given by the Nyquist beam sampling rates

$$Q_1 \geq \frac{4N_1d_1}{\lambda_{\min}}\sin(\frac{\Theta_1}{2})$$

and

$$Q_2 \geq \frac{4N_2d_2}{\lambda_{\min}}\sin(\frac{\Theta_2}{2}).$$

The number of samples cannot be fractional, so Q_1 and Q_2 are typically chosen to be the smallest integer that satisfies these equations. For the wideband pulse-echo ultrasound imaging system, there is also sampling along the axial direction at a temporal sampling rate f_{sample}. As a consequence of the beam and temporal sampling, k-space representations of the resolution such as the PSF are periodic along the lateral spatial frequency axes, with periodicity determined by the beam sampling rates, and along the axial spatial frequency axis, with periodicity determined by the temporal sampling rate f_{sample}. Since the system is sampled, the previously mentioned relationships between the transmit aperture, the receive aperture, the effective aperture, the PSF and the coarray are replaced with their discrete equivalents.

Fig. 6 illustrates the principal components of an ultrasound imaging system **600** including a transducer array **610** and front-end electronics **612** for pulse generation, transmit/receive multiplexing, amplification, filtering, time-gain compensation, digital-to-analog conversion and transmit and receive beamforming. In addition, the ultrasound imaging system **600** contains a controller **614** such as a microprocessor and an image processor **616** including dynamic and static memory such as DRAM and SRAM for beam processing, envelope detection, scan conversion and log compression. The ultrasound imaging system **600** also contains a display **618**. The controller **614** provides all the necessary timing and operation signals to the front-end electronics **612**.

As is known in the art, transmit and receive beamforming are used to vary transmit and receive focal length and transmit and receive direction. Achieving dynamic transmit focusing requires multiple firings from the array **610** for each scan line, while dynamic receive focusing can be performed with only one firing. Since scan time per frame is limited in real-time imaging, transmit focus for FPA imaging is often fixed while receive focusing is performed dynamically. In an alternate embodiment, a plurality of images may be acquired for a plurality of transmit focal lengths. Each scan line is determined by selecting azimuth angle θ_1 **518** and elevation angle θ_2 **520** in **Fig. 5**. Energy in the ultrasound frequency range (2-50 MHz) is transmitting in that direction with a desired focal point by beamforming. Received response is processed by beamforming to shift the receive focal length dynamically. Coherent summation is performed to form the scan line, which has the received response at each receive focal length. A full set of scan lines is obtained by repeating these steps for all beam directions.

Fig. 3 illustrates the process of determining scan lines for FPA imaging for an array **300** that corresponds to a 2D cross-section in the plane defined by the azimuth angle θ_1 **518** or the elevation angle θ_2 **520**. Scan lines **310, 312, 314** up to **316** (corresponding to Q_1 or Q_2) are acquired in sequential steps. Each scan line **310, 312, 314** and **316** has data at receive focal lengths **318, 320, 322, 324, 326** and **328**. As shown previously in **Fig. 1**, for multiple element transducer arrays FPA imaging requires an equivalent number of front-end processing

channels. All elements are used during transmit and receive in order to form each of the Q_1xQ_2 beams.

Referring to **Fig. 5**, the imaging technique of this invention is based on K_1xK_2 subarrays each having multiple adjacent elements M_1 **528** and M_2 **530** in the transducer array **500**. In this technique the subarrays transmit energy with frequencies in the ultrasound region and receive responses to this energy. The elements in active subarray **532** transmit and receive in parallel. Active subarray **532** is multiplexed across the full N_1 **510** x N_2 **514** transducer array **500**.

Since the subarray acquires a smaller range of lateral spatial frequencies than the full array, the beam space sampling requirements are relaxed. It is, therefore, possible to reduce the number of firings without loss of information in the image by sparsely sampling the beam space. The beam-space sampling criterion is dependent only on the active aperture size and not its relative location. To avoid aliasing, the beam sampling rate for PSA imaging with the M_1 **528** x M_2 **530** active subarray **532** is

$$Q_{S1} \geq \frac{4M_1d_1}{\lambda_{\min}}\sin(\frac{\Theta_1}{2})$$

and

$$Q_{S2} \geq \frac{4M_2d_2}{\lambda_{\min}}\sin(\frac{\Theta_2}{2}),$$

where Q_{S1} and Q_{S2} are the number of samples (beams) required in the first lateral direction **512** and the second lateral direction **516** for the active subarray **532**. In this invention, the beam space is coarsely sampled to meet the sampling criteria for each transmit/receive subarray.

The total number of beams for each active subarray **532** is equal to $Q_{S1}xQ_{S2}$. After multiplexing the active subarray **532** over the full transducer array **500**, a total of K_1xK_2 sets of $Q_{S1}xQ_{S2}$ beams are acquired. We refer to these as low-beam-rate subarray images, one for each subarray.

High-beam-rate subarray images correspond to the full set of Q_1xQ_2 beams, and are ideally equal to the images that would have been formed if each of the subarrays had directly acquired all Q_1xQ_2 beams. Increasing the beam density to reconstruct high-beam-rate subarray images from the low-beam-rate subarray images is accomplished by upsampling and interpolating in the planes corresponding to the azimuth angle **518** and the elevation angle **520**. This may be performed in the image processor **616** shown in **Fig. 6**. In an alternate embodiment, upsampling and interpolation is varied for at least some of the subarrays. Typically, the interpolation is accomplished in at least a filter. Alternatively, null scan lines, having zeros for each receive focal length, are inserted between the beams in the low-beam-rate subarray image prior to applying an interpolation filter. Another implementation, is to upsample, apply the interpolation filter and then to discard samples to achieve the desired sample rate. The upsampling factor in the planes corresponding to the azimuth angle **518** and the elevational angle **520** must be chosen as

$$L_1 < \frac{Q_1}{Q_{S1}}$$

and

$$L_2 < \frac{Q_2}{Q_{S2}}.$$

The reconstructed high-beam-rate subarray images are then combined to form the final high-beam-rate PSA image. This step may be accomplished by adding the high-beam-rate subarray image for each subarray to a running summator. In general, this step includes appropriate weighting of each of the high-beam-rate subarray images as well as subarray-dependent spectral modification. All of these steps may be performed in the image processor **616** in **Fig. 6**. With the proper choice of the filters and weights, the quality of the final image is comparable to that achieved using FPA imaging for receive image focal lengths near the transmit focal length. We refer to the imaging technique of this invention as phased subarray (PSA) imaging. It is distinct from CSA, SPA and FPA imaging. Note that the axial resolution is unaffected by PSA imaging.

Referring to **Fig. 6**, the front-end electronics **612** complexity in PSA imaging depends directly upon the subarray size, (M_1 **528** x M_2 **530** channels are required as shown in **Fig. 5**) motivating the choice to make the subarray size much smaller than the full array size, i.e., M_1 **528** much less than N_1 **510** and M_2 **530** much less than N_2 **514**. **Fig. 7** illustrates the front-end transmit **712** and receive electronics **714** of a PSA imaging system **700** with 4 elements **702, 704, 706, 708** in subarray **710** along the first lateral direction **512** or the second lateral direction **516** as defined in **Fig. 5**. Referring to **Fig. 7**, the subarray **710** is driven by drive signals from transmit beamformer **718** via the transmit electronics **712** and multiplexer **716**. Receive signals from the subarray **710** are coupled to receive beamformer **720** via the receive electronics **714** and the mutiplexer **716**. In an alternate embodiment of the invention (not shown), one subarray is used for transmitting energy with frequencies in the ultrasound and a separate subarray is used for receiving the responses to the transmission of this energy.

Referring back to **Fig. 5**, in one embodiment of this invention each subarray has the same size M_1 **528** and M_2 **530** and the subarrays are regularly spaced in first lateral direction **512** and second lateral direction **516**. In another embodiment, each subarray is square, i.e., M_1 **528** and M_2 **530** are equal. Depending on the subarray geometry, the subarrays may overlap with one another in order to allow restoration of all spatial frequencies. Overlap **534** is

defined as adjacent subarrays **532** and **536** having a number of adjacent elements in common. In general, the overlap along lateral directions **512** and **516** need not be the same. If PSA imaging is implemented with the same active subarray **532** for transmit and receive, adjacent, non-overlapping subarrays will cause nulls in the coarray that represent a complete loss of information at those spatial frequencies. Therefore, overlap **534** is required in this embodiment. In an alternate embodiment, where separate subarrays transmit and receive, adjacent and touching subarrays are sufficient to prevent nulls in the coarray.

In addition to the number of samples (beams) that are acquired, the number of firings required for PSA imaging depends on subarray size, M_1 **528** x M_2 **530**, and the number of subarrays, K_1 x K_2, needed to cover the entire transducer array **500** without a loss of information. This provides an additional motivation for keeping the subarray size small. However, to increase the frame rate, which is inversely proportional to the number of firings, the number of subarrays should be kept to a minimum. For a fixed transducer array **500** size and fixed subarray size **532**, decreasing the number of subarrays also implies decreasing the amount of overlap **534** between subarrays.

In general, for an arbitrary amount of overlap **534** (as well as for arbitrary subarray size M_1 **528** and M_2 **530**, arbitrary spacing d_1 **524** and spacing d_2 **526** in the array **500**, and arbitrary subarray and full array **500** aperture functions) the summation of the high-beam-rate subarray images results in an irregularly shaped effective aperture function and is not suitable for imaging. Therefore, one embodiment of this invention includes additional filtering to spectrally modify the high-beam-rate subarray images to produce a more uniform spatial frequency response and to restore the coarray for the final high-beam-rate PSA image to that of an FPA image. This restoration filtering may be combined with the interpolation filter. The step of weighing each of the high-beam-rate subarray images may also be combined with this restoration filtering. In general, the restoration filter is varied for the subarrays. In an alternate embodiment of this invention, the restoration filter is also varied for the receive focal lengths.

The interpolation filtering, and the restoration filtering may be implemented in a single filter or in separate filters. The filter may be analog or digital. For a digital filter, settings for the taps corresponding to filters for different subarrays may be stored in a look-up table. In an alternate embodiment, some of the settings for a digital filter may be calculated using the controller **614** in **Fig. 6** based on values stored in a look-up table (not shown). The filter may be determined by a modification of the optimization technique known in the art in which a stop band weighting function is defined, a pass band weighting function is defined and a multiple-objective weighting parameter is defined. The desired filter is then determined by minimizing the sum of the squared error between desired coarray and reconstructed coarray for all the subarrays and the multiple-objective weighting parameter times the sum of the squared out-of-band (or stop band) energy for all the subarrays.

Since the filter must already be applied to all the low-beam-rate subarray images that are used to form the final high-beam-rate PSA image, additional filtering capabilities may be combined with the filter to address further image enhancement at no extra cost. In an alternate embodiment of the present invention, images may be further enhanced by filtering to correct for temporal spectral imperfections, for defocusing for receive focal lengths outside of the focal zone corresponding to the transmit focal length and to compensation for a non-uniform spatial frequency response.

Images from narrowband systems can use a filter with support only in the lateral directions **512** and **516** in **Fig. 5**, and can have minimal support in the axial direction **522**. Wideband systems also require filter support in the axial direction **522** in order to properly reconstruct high-beam-rate subarray images from their low-beam-rate counterparts. Therefore, in another embodiment, for sufficiently narrowband signals the filter or filters for interpolation and spectral modification (including restoration) is a 1D or 2D filter. In yet another embodiment, for wideband signals the filter or filters for interpolation and spectral modification (including restoration) is a 2D or 3D filter.

Fig. 4 illustrates the process of determining scan lines for PSA imaging for a 2D cross-section in the plane defined by the azimuth angle θ_1 **518** or the elevation angle θ_2 **520** in **Fig. 5**. Referring back to **Fig. 4**, scan lines **410, 412, 414, 416** and **418** (corresponding to Q_{S1} or Q_{S2}) are acquired in sequential steps by active subarray **402** to produce the low-beam-rate subarray image. Each scan line **410, 412, 414, 416** and **418** has data at receive focal lengths **420, 422, 424, 426, 428** and **430**. The active subarray **402** is multiplexed across array **400**. Each subarray acquires one of scan lines **410, 412, 414, 416** and **418** before the next scan line is acquired. In an alternate embodiment, it is possible for each of the subarrays to acquire one of scan lines **410, 412, 414, 416** and **418** consecutively. However, this embodiment is not preferred since it results in a longer time between the acquisition of scan lines **410, 412, 414, 416** and **418** by each of the subarrays and leads to motion artifacts. The low-beam-rate subarray images are then interpolated in a filter. The resulting high-beam-rate subarray images are weighted prior to summation to produce a high-beam-rate PSA image.

The SNR of an array imaging system is dependent upon the number of active transmit and receive channels. Assuming that the noise is additive and statistically independent on the receive channels, the normalized SNR in dB of the PSA imaging system with $M_1 = M_2 = M$ is given by

$$SNR = 20\log\left[M\sqrt{M}\frac{\sum_k b[k]}{\sqrt{\sum_k b^2[k]}} \right] + SNR_o,$$

where b[k] is the weighting of the k-th high-beam-rate subarray image, SNR_0 is the pulse-echo SNR of a single channel, assuming that the SNR of the array channels are identical. Referring to **Fig. 5**, as an example for M_1 **528** x M_2 **530** = 4×4, N_1 **510** x N_2 **514** = 16×16 and $K_1 \times K_2$ = 5×5, the normalized SNR values of the FPA, PSA and SPA imaging systems are 72, 60 and 48

dB, respectively. Table I shows a comparison of the relative theoretical performance of FPA, PSA and SPA imaging systems for square arrays (N_1 **510** = N_2 **514** = N, M_1 **528** = M_2 **530** = M, $K_1 = K_2 = K$, and sector angles $\Theta_1 = \Theta_2 = \Theta$) with no additional restoration or spectral modification filtering besides a bandpass filter. While the SNR performance of PSA imaging is good, in another embodiment high-beam-rate subarray images for active subarray **532** may be acquired a plurality of times and averaged to improve the SNR. In yet another embodiment, low-beam-rate subarray images for active subarray **532** may be acquired a plurality of times and averaged to improve the SNR prior to interpolation.

Table 1. Performance comparison of FPA, PSA and SPA imaging.

	Number of Firings	
Algorithm	**Exact**	**Numerical Example (N=32, M=8, K=7, Θ=90°, d=λ_{min}/2)**
SPA	N^4	1048576
FPA	$[\frac{4Nd}{\lambda_{min}}\sin(\frac{\Theta}{2})]^2$	2048
PSA	$[\frac{4KMd}{\lambda_{min}}\sin(\frac{\Theta}{2})]^2$	6272
	Front-end Hardware Complexity	
Algorithm	**Exact**	**Numerical Example (N=32, M=8, K=7, Θ=90°, d=λ_{min}/2)**
SPA	1	1
FPA	N^2	1024
PSA	M^2	64
	Signal-to-Noise Ratio (SNR) in dB	
Algorithm	**Exact**	**Numerical Example (N=32, M=8, K=7, Θ=90°, d=λ_{min}/2)**
SPA	$20\log_{10}(N^2)$	90
FPA	$20\log_{10}(N^3)$	90
PSA	$20\log_{10}\left[M^3(\frac{\sum_k b[k]}{\sqrt{\sum_k b^2[k]}})^2\right]$	69

Example 1

For the special case of a fixed number of adjacent elements in each subarray and a fixed overlap **534** in **Fig. 5** equal to half the number of adjacent elements M_1 **528** or M_2 **530** along lateral directions **512** or **516** in each subarray, the filter for interpolation is merely a bandpass; no additional spectral modification in a restoration filter is required. For narrowband imaging, the bandpass is 1D. For wideband imaging, the bandpass is in general 2D. The final high-beam-rate PSA image is a linear combination of individual high-beam-rate subarray images. For a 2D cross-section in the plane defined by the azimuth angle θ_1 **518** or the elevation angle θ_2 **520**, the weight b[k] applied to the high-beam-rate subarray image corresponding to k^{th} subarray is given by

$$b_1[k] = (\frac{K_1+1}{2}) - \left|k - (\frac{K_1-1}{2})\right|,$$

where k is between 0 and $K_1 - 1$. For this geometry, and more generally for overlap **534** less than half the number of adjacent elements M_1 **528** or M_2 **530**, the frame rate reduction will never exceed a factor 2 in 2D and a factor of 4 in 3D.

Fig. 10a-c illustrates the 2D lateral and axial spatial frequency response for PSA imaging corresponding to a cross-section of 3D data in the plane defined by the azimuth angle θ_1 **518** or elevation angle θ_2 **520** in **Fig. 5**. For convenience, take the plane defined by the azimuth angle θ_1 **518**. In this example, $K_1 = 7$ subarrays span all N_1 **510** elements in **Fig. 5**. Each subarray is composed of M_1 **528** = $0.25xN_1$ **510** elements. The subarrays overlap **534** by $0.125xN_1$ **510** elements. The number of beams in the subarray images is equal to one-third the number of beams in the final image ($Q_1/Q_{S1} = 3$).

The horizontal and vertical axes in **Fig. 10a-c** are the lateral and temporal frequency, respectively. Note that the axial spatial frequency and the temporal frequency are related by the speed of sound. The projection of the spatial frequency response onto the temporal

frequency axis is equal to the Fourier transform of the pulse response. Note that the spatial frequency response in **Fig. 10a-c** for subarrays that are not at the center of the array **500** in **Fig. 5** are sheared, where the shear factor is proportional to the distance of the respective subarray from the center of the array.

The process of converting the real received signal to a complex-valued analytic signal eliminates all negative temporal frequency components of the signal. The effect on k-space is that there is no signal contribution for temporal frequency less than zero. Therefore, the corresponding bottom half of the illustration in **Fig. 10a-c** is not shown.

Fig10a-c shows the theoretical nonzero portions of the 2D spatial frequency response at each stage of the image acquisition and formation process. The first step is to acquire Q_S beams from each of the K_1 subarrays. By reducing the number of directly acquired beams the frame rate is only reduced by less than a factor of 2. Two spatial frequency responses for these low-beam-rate subarray imaging systems are shown in **Fig. 10a**. The top illustration corresponds to a subarray at the center of the array and the lower illustration to an off-center subarray. The lateral frequency width **1000** of both spatial frequency response is $2M_1d_1/\lambda$. The next step is to upsample these images by inserting zero-valued beams between the acquired beams. The total number of beams in the upsampled images is the same as that in the FPA system, i.e., Q_1. The spatial frequency response response after upsampling is shown in **Fig. 10b**. These spatial frequency responses represent the periodic replication of the low-beam-rate spatial frequency responses shown in **Fig. 10a**. In this example, the high-beam-rate subarray images are obtained by applying a bandpass filter. The passband of such a filter is shown by the striped background in **Fig. 10b**. This filter can be applied by convolution in the spatial domain or by multiplication in the spatial frequency response domain. The filter effectively suppresses the replicas of the original spatial frequency response response; no restoration filter is required. After coherent weighting and summation of all 7 high-beam-rate subarray images, the spatial frequency response response becomes comparable to that of FPA imaging (see **Fig. 10c**) with a lateral frequency width **1010** equal to $2N_1d_1/\lambda$.

Example 2

Fig. 8 illustrates the impact of restoration filters including the subarray weights on the high-beam-rate PSA coarray. **Fig. 8** shows the coarray corresponding to a 1D lateral cross-section of 3D data in the plane defined by the azimuth angle θ_1 **518** or elevation angle θ_2 **520.** In this example, N_1 **510** or N_2 **514** = 10, M_1 **528** or M_2 **530** = 6 and K_1 or K_2 = 3 in **Fig. 5**. Referring back to **Fig. 8**, coarrays **810, 812** and **814** each correspond to a subarray. Each coarray **810, 812** and **814** has 11 non-zero samples (2 M_1 **528** -1 or 2M_2 **530** -1). Without restoration filtering, the weighted sum of coarrays **810, 812** and **814** results in an unrestored PSA coarray (not shown) that is not suitable for producing a high-beam-rate image. One possible set of restoration filters **816, 818** and **820** that could be used to obtain a high-beam-rate PSA coarray **828** that is comparable to a FPA coarray are shown. In this illustration, all the weights are equal to 2, and are incorporated into the magnitudes of the restoration filters **816, 818** and **820**. The products of the coarrays **810, 812** and **814** with the restoration filters **816, 818** and **820** produce restored coarrays **822, 824** and **826**. The sum of these restored coarrays **822, 824** and **826** is the desired high-beam-rate PSA coarray **828** in the plane defined by the azimuth angle θ_1 **518** or elevation angle θ_2 **520** in **Fig. 5**. PSA coarray **828** has 19 non-zero samples (2 N_1 **510** -1 or 2N_2 **514** -1)

Example 3

A comatrix serves as a useful tool for choosing which transmit and receive subarrays should be used to form the final coarray. Two example comatrices are shown in **Fig. 9a-b**, all with N_1 **510** or N_2 **514** = 16 and M_1 **528** or M_2 **530** = 4 (once again, **Fig. 9a-b** illustrates a 2D cross-section in the plane defined by the azimuth angle θ_1 **518** or elevation angle θ_2 **520** in **Fig. 5**). Each example demonstrates how different choices for the number of subarrays and the transmit/receive subarray combinations affect the restoration filter needed to achieve an FPA-comparable coarray.

For the example shown in **Fig. 9a**, the array **900** is divided into four non-overlapping, adjacent subarrays **910**, **912**, **914** and **916**. Images are acquired using 16 permutations **918**, **920**, **922**, **924**, **926**, **928**, **930**, **932**, **934**, **936**, **938**, **940**, **942**, **944**, **946** and **948** of the subarrays **910**, **912**, **914** and **916**, one for every transmit/receive combination. The weights used in summing the coarrays for permutations **918**, **920**, **922**, **924**, **926**, **928**, **930**, **932**, **934**, **936**, **938**, **940**, **942**, **944**, **946** and **948** are all 1, i.e., weighting is not required in this configuration. The comatrix is fully-populated and equivalent to the comatrix for FPA or SPA imaging. In this configuration, the resulting coarray (not shown) is comparable to that of FPA without any restoration filtering. The disadvantage of this configuration is that each beam must be acquired 16 times. The frame rate can be improved by reducing the number of active transmit/receive subarray combinations.

For the example shown in **Fig. 9b**, the same subarrays **910**, **912**, **914** and **916** are employed, but only 10 transmit/receive subarray combinations are used (**918**, **920**, **926**, **928**, **930**, **936**, **938**, **940**, **946** and **948**). Combining the coarrays without filtering or weights would result in non-uniform coarray (not shown). Fortunately, the desired FPA-comparable coarray is a linear combination of the coarrays, with weights of 1, 1, 1, 3, 2, 2, 3, 1, 1 and 1 for combinations **918**, **920**, **926**, **928**, **930**, **936**, **938**, **940**, **946** and **948**, respectively.

Other geometries with overlap **534** and different permutations of transmit/receive combinations (not shown) reduce the number of firing events per beam required to obtain the coarray. For each geometry and the transmit/receive combinations selected in the comatrix, different weights are required to obtain the FPA-comparable coarray. Some geometries with overlap **534** require restoration filtering to obtain the FPA-comparable coarray. If N_1 **510** or N_2 **514** = 16, M_1 **528** or M_2 **530** = 4, K_1 or K_2 = 7, overlap **534** = 2 (half of M_1 **528** or M_2 **530**) and the same subarray is used to transmit and receive, no restoration filtering is required (this example corresponds to the conditions in Example 1). In this case, only one firing is performed per subarray, since each subarray is acting as both transmitter and receiver. The appropriate weights starting at one side of the array **900** and moving to the other side are 1, 2,

3, 4, 3, 2 and 1. If N_1 **510** or N_2 **514** = 16, M_1 **528** or M_2 **530** = 4, K_1 or K_2 = 5, overlap **534** = 1 and the same subarray is used to transmit and receive, restoration filtering is required to reshape the coarrays in such a way that they can be linearly combined to form the desired final FPA-comparable coarray. Thus, the number of firings per beam was decreased further by reducing the amount of overlap **534** between adjacent subarrays at the expense of requiring restoration filtering.

Example 4

Additional analysis and measurements of PSA imaging have been performed. In the analysis and measurements:

- N_1 **510** = 128 and N_2 **514** = 1;
- spacing d_1 **524** (250 microns in the measurements) is equal to half of the minimum wavelength;
- in the analysis, the temporal sampling frequency was equal to 96 MHz, the pulse center frequency was 3 MHz and, after beamforming, the temporal sampling rate was 12 MHz;
- in the measurements, the temporal sampling frequency was 50 MHz, the pulse center frequency was 3 MHz and, after beamforming, the temporal sampling rate was 12 MHz;
- 80% signal bandwidth;
- M_1 **528** = 32;
- K_1 = 7;
- Q_{S1} equal to 127 beams;
- Q equal to 511;
- transmit focal length in the measurements of 13 cm;
- sector angle Θ_1 in the measurements of 90°;
- and the reconstruction filter was 2D with 31x31 taps.

The experimental data was acquired using a capacitive micromachined ultrasound transducer array imaging several thin wires placed in vegetable oil. The experimental set-up is further described in Ö. Oralkan et al, "Capacitive Micromachined Ultrasonic Transducers: Next Generation Arrays for Acoustic Imaging?," IEEE Trans. Ultrason., Ferroelect., Freq. Contr., vol. 49, pp. 1596-1610 (2002). The analysis and experimental results are further described in Jeremy A. Johnson et al., "Coherent Array Imaging Using Phased Subarrays — Part I: Basic Principles," preprint (2003) and Jeremy A. Johnson et al., "Coherent Array Imaging Using Phased Subarrays — Part I: Simulations and Experimental Results," preprint (2003), the contents of which are incorporated by reference. Good agreement between simulated and measured PSF and corray at each step in PSA imaging was found. In the experiments, the resultant high-beam-rate PSA B-scan images of a phantom wire were in good agreement with the FPA B-scan images.

Both the simulated and experimental results demonstrate the success of the invention for a particular choice of imaging parameters. In this case, the number of dedicated front-end hardware channels needed for both transmit and receive was reduced by a factor of 4, from 128 to 32. Compared to FPA imaging, this example decreases the frame rate by 43%.

PSA imaging has largely been described for the case of 3D imaging with a 2D transducer array. However, as illustrated in Example 4, PSA imaging may be applied to 2D ultrasonic imaging with a 1D transducer array with N_2 **514** in **Fig. 5** equal to 1. In this embodiment, for sufficiently narrowband signals the filter or filters for interpolation and spectral modification (including restoration) is a 1D filter. For wideband signals in this embodiment, the filter or filters for interpolation and spectral modification (including restoration) is a 2D filter.

In ultrasound imaging, the received ultrasound signals are amplitude modulated about a carrier. In one embodiment of this invention, high-beam-rate PSA imaging is applied to the received signals before converting to baseband. In an alternative embodiment of this invention,

the received signals are converted to baseband before high-beam-rate PSA imaging is applied. In yet another alternative embodiment of this invention, the received signals are converted to an intermediate frequency before high-beam-rate PSA imaging is applied.

The high-beam-rate PSA image method may be applied to real-valued receive signals. In this embodiment, the filters employed in PSA imaging are, in general, complex. In another embodiment, a Hilbert transform is applied to the real-valued receive signals to generate complex signals and the filters employed in PSA imaging are real.

The method can also be applied to many variations that have not been described here, such as uneven spacing d_1 **524** and spacing d_2 **526** in **Fig. 5**, non-uniform beam sampling as well as non-uniform or apodized apertures.

The above description has assumed that a transducer array **500** in **Fig. 5** was used for forming images from acoustic waves. However, the theory applies to any coherent array imaging system, and is applicable to areas such as radar, optics, sonar, radio astronomy, seismic imaging, and other imaging modalities.

In view of the above, it will be clear to one skilled in the art that the above embodiments may be altered in many ways without departing from the scope of the invention. Accordingly, the scope of the invention should be determined by the following claims and their legal equivalents.